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Beyond the classic correction system: a numerical nonrigid approach to the scoliosis brace

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BACKGROUND CONTEXT: Adolescent idiopathic scoliosis (AIS) causes a spine and rib cage three-dimensional (3D) deformity previously treated by bracing. Whatever the manufacturing pro-cess, this rigid system acts biomechanically on the patient through the ''three-point bending'' mechanical principle. It applies corrective forces to a limited area and acts especially in the frontal plane. It seemed to us that a nonrigid system, called ''Cbrace,'' with 3D action allowing distribution of forces could increase compliance and provide better long-term correction prospects.

PURPOSE: The aim of this study was to design a nonrigid brace by numerically testing in a finite-element model developed here.

STUDY DESIGN: A finite-element model has been developed to simulate brace effect on AIS right thoracic deformation of a 10-year-old patient.

METHODS: A two-step method was needed; first, the reliability of our model is evaluated, and then, the ability to use distributed forces to correct scoliosis deformation is tested. To obtain a 3D correction, several treatments are experimented, leading to a comparison test between the best combination to the ''three-point bending'' principle.

RESULTS: The numerical model developed here shows good qualitative answers for the treatment of brace forces. The first results demonstrate numerically that distributed forces may be of interest in brace treatment design. Overall force of 40 N above cartilage of the last nonfloating ribs asso-ciated to two posterior asymmetrical areas appears to be the best way to correct scoliosis deforma-tion with nonrigid action.

CONCLUSION: The results show numerical efficacy of distributed forces to correct spinal defor-mities and raises the prospect that a new numerical brace, called ''Cbrace,'' could be a starting point in the search for a nonrigid system.

Introduction

Adolescent idiopathic scoliosis (AIS), the most common type of scoliosis, is a musculoskeletal disease that causes a spine and rib cage three-dimensional (3D) deformity [1]. Although surgical intervention can reduce it, the main clinical goal of orthopedic or physiotherapy treatment is to prevent spinal deformity from increasing. This is the greatest problem in scoliosis treatment [2], and a corrective brace is usually suggested. Although designed in different manners, they are still based on dated concepts, proposed and tested decades ago when the 3D nature of AIS was rarely considered or incorporated into brace design.

The Boston Brace System is one of the most widely used types of thoracolumbosacral orthosis (TLSO) in North America to correct Cobb angle (frontal deformity) and to work toward reducing gibbosity (horizontal deformity), and the Cheneau-Toulouse-Munster is its equivalent in

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Europe. Both act to decrease frontal deformity but do not reestablish physiological thoracic sagittal curves [3,4]; they reduce spinal deformity through a ''three-point bending'' mechanical action, applying corrective forces to a limited area (single-point forces). However, this brace treatment falls short of its goals because of its limited effect on the plane of maximum deformity; moreover, it has no significant derotational effect on the horizontal plane [3]. Furthermore, these rigid systems of correction decrease children's quality of life [5], leading to poor compliance and disturbed bone growth. Thus, they are not an optimal long-term solution.

However, before clinical experimentations, it would seem judicious to perform numerical tests evaluating the chances of successful design of a bracing system, thus analyzing the effect and influence of a single component within the construct investigated. Some specific models of the spine and rib cage use time increments [6,7], simulating growth to mirror the scoliosis mechanism; others [8–11], including ligamentoskeletal models, aim at a quantitative evaluation of brace treatment or optimize the location of TLSO brace action, but none of them have been used to provide a new strategy of scoliosis treatment.

An examination of the anatomy of thoracic vertebrae reveals their high capacity for horizontal rotation in hypokyphosis (80% in thoracic scoliosis); that is why, to avoid the 3D deformity increasing, a promising way to decrease this rotational potential is to reestablish physiological sagittal spinal curves in the thoracic area. One possible hypothesis is that instead of single-point force, a nonrigid system using straps allowing distribution of forces could yield a more flexible brace able to increase compliance and offer better long-term correction prospects.

The aim of this study was to numerically evaluate mechanical effects of a potential nonrigid brace, called ''Cbrace,'' in a finite-element model developed here. This concept of the nonrigid brace is intended to correct a typical spine and rib cage deformity of a standard right thoracic scoliosis. The study is numerical, is limited to a single-curve type, and evaluates only the immediate effect of this new brace system. The reliability of this modeling of the ''virtual patient'' is first evaluated, and then, the efficacy of distributed forces to correct scoliosis deformity is tested. Finally, to obtain ''Cbrace,'' several treatments are tested and compared with the classic treatment.

Method

General

A two-step method for nonrigid brace design is used. First, a finite-element modeling of a ''virtual patient'' is built, and second, several numerical simulations are performed in three parts. First, correction via distributed forces is tested; second, several spine corrections are evaluated in each dimension; and third, the most efficient combination of the previous treatment is compared with the classic.

Step 1. Model building and evaluation

The ''virtual patient'' is a 10-year-old (with Risser Level 0) with a ''regular'' right thoracic scoliosis with a risk of progression of 95% [12] that requires brace treatment, the ideal configuration for a clinical research protocol [13].

The model includes bones and intervertebral discs (IVDs). Muscles were not considered in the model as they have not been proven to actively contribute to brace correction [14]. Modeling software Hypermesh (Altair Hyperworks, Troy, MI, USA) was used for building and meshing the geometry of the scoliosis. For simulation of the brace treatment, Abaqus finite-element software was used (Simulia, Dassault Systems, Velizy Villacoublay Cedex, France). These softwares are general displacement-based finite-element codes and used to calculate the 3D nodal displacement (Fig. 1).

Geometric parameters

The spine size and the bone morphology (as rib cage, vertebrae, sternum, and cartilage) were obtained by one X-ray scoliosis imaging (CHU Marseille-Timone) of a 10-year-old girl before puberty (Risser Level 0). Rib curves in the sagittal plane and rib-cartilage curves (7th to 10th) were taken from Stokes et al. [15], and distribution of vertebra rotation in the horizontal plane was taken from Berthonnaud and Dimnet [16]. In this model, the equivalence of a horizontal rotation is a rib rotation (maximum of 8° at the apex level). The different angle values are shown in Table 1. The spine is composed of bones and IVDs, and the rib cage is composed of bones and cartilage. All rib-sternum and rib-vertebra joints were considered as embedded.

Components model

To imitate the geometry of a scoliosis deformity (spine and rib cage), the model includes the following:

- 17 vertebrae (5 lumbar and 12 thoracic): A meshing disc of hexahedral elements is extruded to reach the height of each vertebral corpus. The diameter of each disc is twice the corresponding vertebral corpus (to take into account the *spinous process*).
- 10 ribs (floating ribs were excluded from the model): An extrusion line of eight hexahedral elements extends from the vertebra to the sternum. This line follows a curve plotting the posterior rib arch.
- The sternum is a hexagonal volume of the rib thickness, meshed with hexahedral elements.
- Sagittal and frontal spine curves are modeled through differences in IVD height. The finite-element model contains 90,406 nodes, 89,361 free degrees, and 70,304 mesh elements.

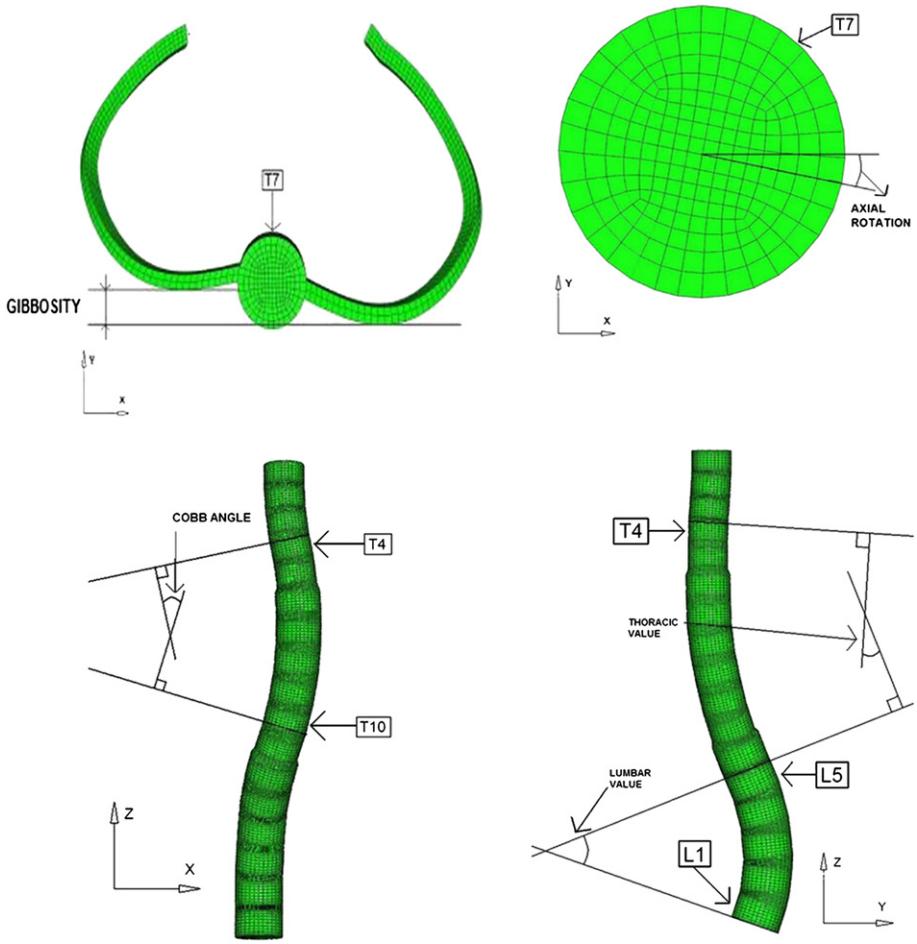


Fig. 1. Parameters evaluated.

Boundary conditions and mechanical properties

To simulate eye stabilization, displacements are allowed only in the *Z* direction of the top of T1 (thoracic vertebra 1) and to simulate brace restraint on the pelvis on the bottom of L5 (lumbar vertebra 5), no displacement is allowed (embedded). The model requires 5 degrees of freedom to be fixed to reach a satisfactory solution. It is a finite-element modeling technique using isotropic linear mechanical properties, taken from experimental and published data on cadaveric spines by Descrimes et al. [8] (Table 2). The method of calculation is based on the Hook law for isotropic materials.

Preliminary evaluation of the "virtual patient"

The simulation of Gignac et al. [17] was used to test the TLSO action described in Fig. 2:

1. Limited force of 40 N applied at the apex level on anterior gibbosity.
2. Limited force of 40 N applied on convex curve.
3. Reactive force on gibbosity.

Step 2. Three-part numerical simulation protocol

The numerical protocol first tests the efficacy of straps in correcting scoliosis deformity; these straps distribute forces instead of applying forces to a limited area, as with the TLSO brace. An overall corrective force of 40 N [18] is applied with 6-cm–wide straps in all the treatments (Fig. 3). The second part of the simulation seeks an optimization of straps' position to correct spinal deformity in three dimensions; all positions were evaluated in each plane. Finally, the third part compares the efficacy of a combination of these straps' positions with the classic (TLSO configuration) and control (model without correction).

The treatment objective is to decrease Cobb angle (frontal deformity in degrees), gibbosity (horizontal deformity in millimeters), and horizontal rotation at the thoracic apical level (horizontal deformity in degrees) and to increase thoracic angle.

Part 1: evaluation of distributed forces

A classic TLSO correction (three single-point forces), called A, is compared with "distributed forces" (B), also

Table 1
Scoliosis values

	Values of scoliosis	
Levels	Clinical data	Numerical
Cobb angle (°)	25	24.4
Lumbar lordosis (°)	40	40.9
Kyphosis (°)	20	20
Gibbosity (mm)	14	13.6

Table 2
Mechanical properties

Elements	Young modulus (MPa)	Poisson ratio
Ribs	5,000	0.1
Sternum	10,000	0.2
Vertebrae	1,000	0.3
IVDs	15	0.45
Rib cartilage	480	0.1

IVDs, intervertebral discs.

representing a TLSO configuration but applied over a large area. B is composed of Treatment 1 (a pair of straps, front/back location) to reduce horizontal deformity and Treatment 2 (one lateral strap on convex curve) to reduce frontal deformity.

Part 2: evaluation of several treatments per correction to optimize 3D action

- Frontal correction

  Treatment 3: area on lateral part of thorax with horizontal direction.

- Sagittal and horizontal corrections:

  Treatment 4: areas horizontally above bone cartilage (last nonfloating rib) in variable directions: Z, $-\cos(40°)Y+\sin(40°)Z$, or $-\cos(65°)Y+\sin(65°)Z$.
  Treatment 5: areas horizontally above cartilage bone on last left rib with direction: $\cos(65°)Y+\sin(65°)$.

- Horizontal correction

  Treatment 6: one or two posterior (asymmetrical) areas in Y direction.

- Three-dimensional action

  Treatment 7: posterior harness above the top of posterior gibbosity.
  Treatment 8: forces distributed on left shoulder (direction $-Z$).
  Treatment 9: anterior harness crossing on sternum (force direction $-Y$).

Part 3: evaluation of a 3D action

The most effective combination from the above treatments (Part 2), named "C" is compared with both the initial and the TLSO configurations with distributed forces (B).

Results

For the same intensity, position, and mechanical properties, the results of Step 1 show similar corrective trends in Cobb angle reduction, gibbosity decrease, and action on sagittal angle (Fig. 4). Our model can be used to compare clinical indices for brace simulations, and qualitative data can be analyzed. Calculation time of our model is less than 1 hour.

Results for the first part of Step 2 show a downward trend in gibbosity and Cobb angle for both treatments, but the reduction is sharper for the rib hump when the corrective forces are distributed (7 mm instead of 9.5 mm). Treatments 1 and 2 meet their own objectives of correction, but neither treatment obtains a specific impact on sagittal curve.

In the second part, to reduce frontal deformity, Treatment 3 produces similar decreases in Cobb angle to Treatment 2 (differing in strap direction) without impact on sagittal curves. For sagittal and horizontal corrections, neither Treatment 4 nor Treatment 5 is able to reduce gibbosity and increase sagittal curve at the same time without squashing the rib cage. To prevent harm to the rib cage, less intense distributed forces (20 N) were applied; the same trends are observed, with slighter rib cage deformities. Treatment 6 achieves its goal of reducing the horizontal deformity specifically (rib hump: 10.2 mm). To reduce

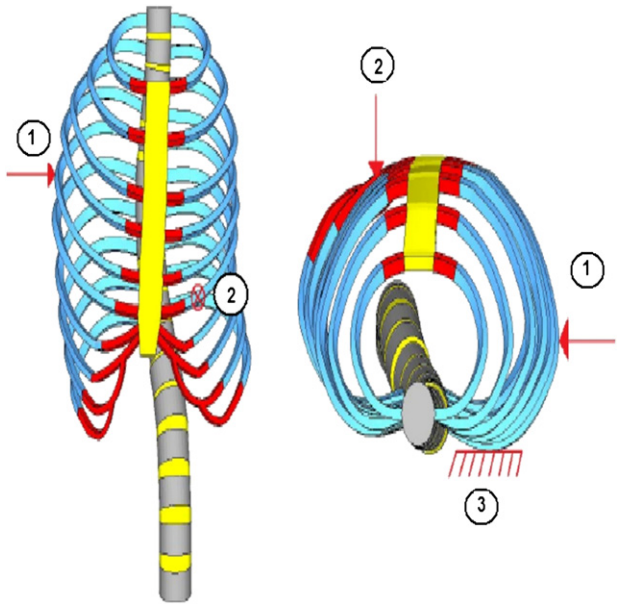


Fig. 2. Evaluation.

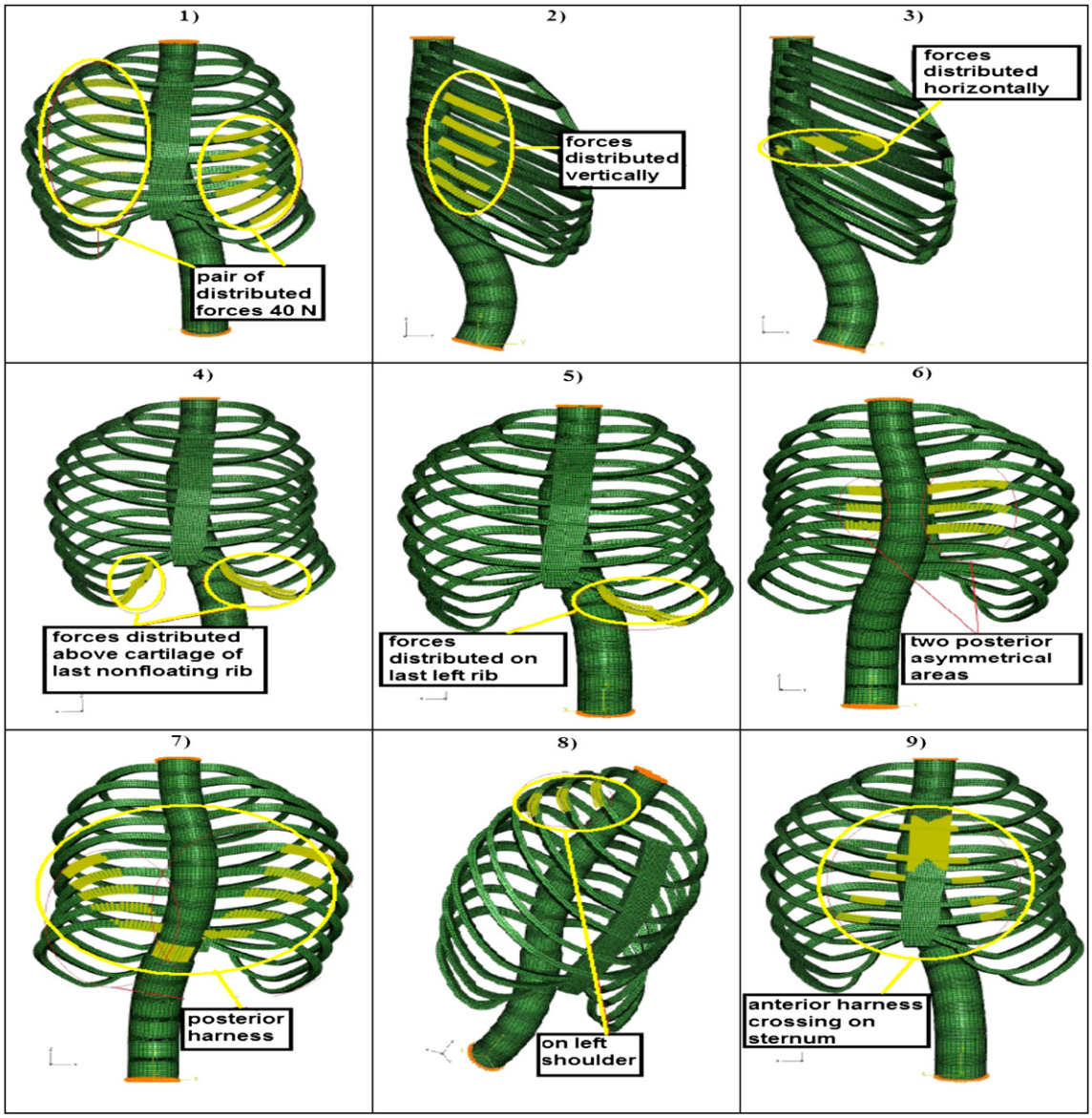


Fig. 3. Treatments.

overall deformity, Treatment 7 (back harness) shows no impact on Cobb angle value but decreases gibbosity (less than 10 mm) and leads to decreased sagittal curves on thoracic and lumbar levels. Its opposite, Treatment 9 (anterior harness), shows no impact on Cobb angle, increases gibbosity (14.5 mm), and increases sagittal curves. Treatment 8 decreases Cobb angle (23.5°) and has no impact on horizontal rotation but leads to a trend of increased sagittal curves, particularly at the thoracic level (22.5°).

In the third part, the most effective combination of these previous treatments (C) (Fig. 5) was obtained by combining Treatment 6: two posterior asymmetrical areas in Y direction, and Treatment 4: large areas distributing force horizontally above bone cartilage (last nonfloating rib) in $-$COS (40°) Y+SIN (40°) Z directions. C was compared with the TLSO configuration with straps (combination B) and with a control (model without correction). The results show a similar trend in clinical parameters for both. C qualitatively increases sagittal curves in the thoracic area, a factor in preventing increasing spinal deformity (Fig. 6); there is a limited impact on lumbar values. These results meet our overall treatment objective.

Discussion

Results from Step 1 showing a qualitatively similar trend to that obtained by Gignac et al. [17] indicate that the

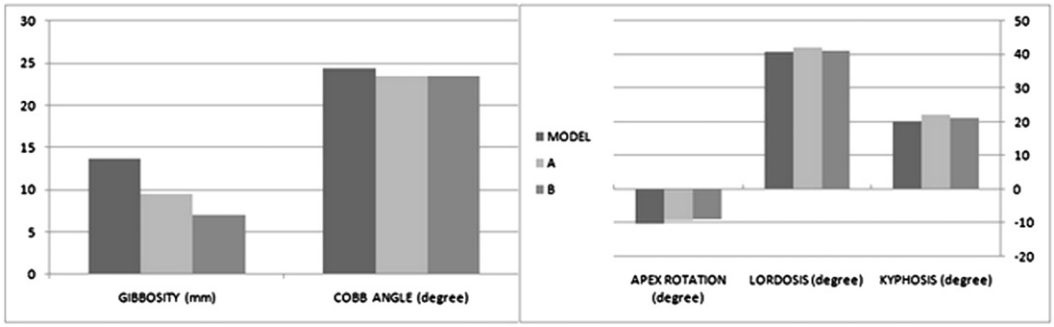


Fig. 4. Evaluation results.

model presented here can be used to design an alternative brace treatment for right thoracic scoliosis. When distributed corrective forces and single-point forces are compared in Step 2, the results show similar reductions for both approaches (A and B) in Cobb angle, gibbosity, and sagittal angulation. Using a nonrigid brace design with straps, it is thus numerically demonstrated that straps inducing distributed forces (B) may be of clinical interest because they lead to improved patient comfort and compliance.

Findings from Part 2 of the second step suggest that the same results can be obtained for different directions (Treatments 2 and 3) using lateral straps, offering a flexibility in brace design. Furthermore, because Treatments 4 and 5 scratched the cage, a decrease in intensity (20 N) was tested for each strap. However, because our model yields only trends in correction, and results must be compared in the same conditions, this optimization method was rejected.

To prevent the common increase in deformity, thoracic angles need to be corrected to physiological values, especially for thoracic angles of more than 20°. The present 3D analysis enables different treatments' action on sagittal thoracic curves to be compared. Thus, from an analysis of several treatments (Treatments 4–9), the combination of Treatments 4 and 6 appears the best way to correct frontal horizontal deformities (Cobb angle and gibbosity) and sagittal thoracic values simultaneously. This most effective combination is called (C).

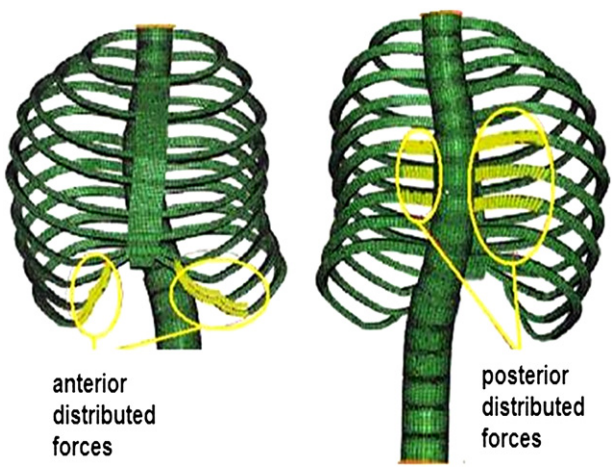


Fig. 5. The most effective combination (C).

The comparison of (C) and (B) shows similar numerical trends for the scoliosis model; (C) appears as a numerical equivalent to the TLSO brace. Even if some authors show only a slight difference in effect between part-time (12–16 hours) and full-time (23 hours) wearing of a brace [19–22], the usual rigid brace design leads to poor adolescent compliance, a major cause of treatment failure. The (C) combination, unlike either the classic rigid Boston or Cheneau-Toulouse-Munster braces [23] or nonrigid brace like Trial C [24], leaves the breast development zone and the scapula area free to move. Using the (C) nonrigid numerically defined strap brace concept: ''Cbrace'' (Figs. 7 and 8) acts on single-curve deformity, and its design could allow trunk movement and offer a greater patient comfort. To prove a better compliance, after development step, a clinical investigation is needed.

In the Society on Scoliosis Orthopaedic and Rehabilitation Treatment consensus paper on brace action [25], a force vector pushing the thoracic convexity was recommended by most participants, together with a ''pair of forces'' to derotate in the axial plane and a coupling mechanism to increase thoracic kyphosis. ''Cbrace,'' applying distributed forces on the anterior part of the last nonfloating ribs and an asymmetric pair of distributed forces on each side of the spine at the apex level, reaches this goal numerically and goes some way toward reestablishing physiological sagittal thoracic curve. But according to some reports on the effect of a change in sagittal alignment, better brace correction can be provided in scoliosis by ''lumbar flexion (delordosation)'' [26,27]. The model presented here has no pelvis and lower segment and, therefore, has no notable impact on lumbar segment; this would be a useful avenue for future research.

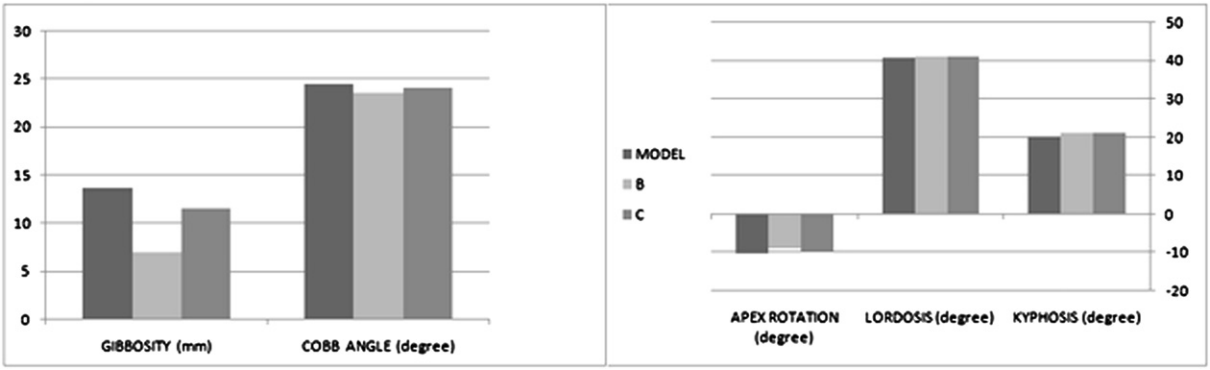


Fig. 6. Optimization results.

Furthermore, it would be valuable to test TLSO with axillary extension (prescribed in cases of AIS with an apex between T7 and T9); to do so, the model used here should be extended to shoulder and scapula.

Because material and geometric linearity in the model have been assumed, these results are only applicable to the situations in which these assumptions apply. The model shows the immediate effect of treatment and does not take into account long-term relaxation of elements and the passive state of muscles. Because little is known about the mechanism of force transmission from abdomen surface to spine, further investigations need to be performed to analyze this transfer. Furthermore, given the vital importance of the IVDs in scoliosis, an interesting avenue of future research could be an analysis of IVD sensitivity during simulations.

Conclusion

Apart from the ''three-point bending system,'' no corrective mechanical principle seems to be universally accepted. The finite-element model used here has shown its reliability in comparing different corrective force applications. Our study is only numerical and limited to a single curve, but it is the first to numerically test distributed forces to correct spinal deformities. Our results raise the prospect that ''Cbrace'' could be a starting point in the search for a new principle of nonrigid brace treatment. Because it is numerically proven to correct spinal deformity without disturbing parts of the trunk and can be designed with nonrigid materials, ''Cbrace'' may offer greater efficacy through allowing trunk movement, thereby improving adolescent compliance.

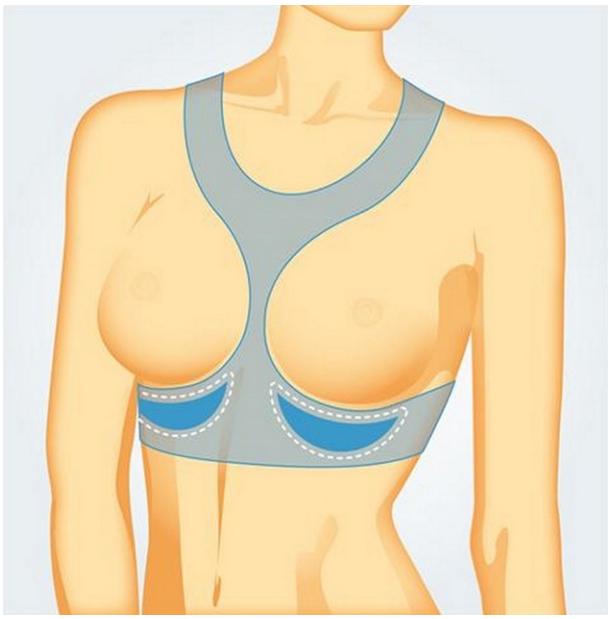


Fig. 7. Cbrace (front view).

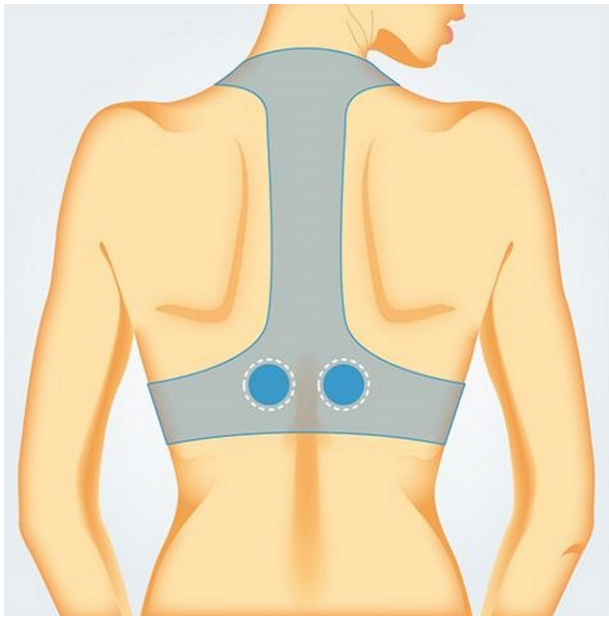


Fig. 8. Cbrace (back view).

References

[1] Panagitis K, Christos K, Grigoris P, Soucacos PN. Effects of thoracolumbosacral orthosis on spinal deformities, trunk asymmetry, and frontal lower rib cage in adolescent idiopathic scoliosis. Spine 2000;25:2064–71.

[2] Castro FP Jr. Adolescent idiopathic scoliosis, bracing, and the Hueter-Volkmann principle. Spine J 2003;3:180–5.

[3] Labelle H, Dansereau J, Bellefleur C, Poitras B. Three-dimensional effect of the Boston brace on the thoracic spine and rib cage. Spine 1996;21:59–64.

[4] Aaro S, Bustrom R, Dahlbom M. The derotating effect of the Boston brace: a comparison between computer tomography and a conventional method. Spine 1981;6:477–82.

[5] Ugwonali O, Lomas G, Choe J, et al. Effect of bracing on the quality of life of adolescents with idiopathic scoliosis. Spine J 2004;4: 254–60.

[6] Stokes IAF, Laible JP. Three-dimensional osseo-ligamentous model of the thorax representing initiation of scoliosis by asymmetric growth. J Biomech 1990;23:589–95.

[7] Huynh AM, Aubin CE, Mathieu PA, et al. Simulation of progressive spinal deformities in Duchenne muscular dystrophy using a biomechanical model integrating muscles and vertebral growth modulation. Clin Biomech (Bristol, Avon) 2007;22:392–9.

[8] Descrimes JL, Aubin CE, Skalli W, et al. Modelling of facet joints in a finite element model of the scoliotic spine and thorax: mechanical aspects. Rachis 1995;7:301–14.

[9] Andriacchi T, Schultz A, Belytschko T, et al. A model for studies of mechanic interactions between the human spine and rib cage. J Biomech 1974;7:497–507.

[10] Closkey RF, Schlutz AB, Luches CW. A model for studies of the deformable rib cage. J Biomech 1992;25:529–39.

[11] Wynarsky GT, Schultz AB. Optimization of skeletal configuration: studies of scoliosis correction biomechanics. J Biomech 1991;24: 721–32.

[12] Duval-Beaupère G. Threshold values for supine and standing Cobb angles and rib hump measurements: prognostic factors for scoliosis. Eur Spine J 1996;5:79–84.

[13] Richards BS, Bernstein RM, D'Amato CR, Thompson GH. Standardization of criteria for adolescent idiopathic scoliosis brace studies: SRS Committee on Bracing and Nonoperative Management. Spine 2005;30:2068–75.

[14] Wynarsky GT, Schultz AB. Trunk muscle activities in braced scoliosis patients. Spine 1989;14:1283–6.

[15] Stokes I, Dansereau J, Moreland MS. Rib cage asymmetry in scoliosis. J Orthop Res 1989;7:599–606.

[16] Berthonnaud E, Dimnet J. Fast calculation of parameters of scoliosis in frontal view for clinical applications. ITBM-RBM 2006;27:56–66.

[17] Gignac D, Aubin CE, Dansereau J, et al. A biomechanical study of new orthotic treatment approaches for the 3D correction of scoliosis. [in French]. Ann Chir 1998;52:795–800.

[18] Petit Y, Aubin CÉ, Dansereau J, et al. Effect of strap tension on the pressure generated by the Boston brace on idiopathic scoliosis patients: a preliminary study. J Biomech 1998;31:175.

[19] DiRaimondo CV, Green NE. Brace-wear compliance in patients with adolescent idiopathic scoliosis. J Pediatr Orthop 1988;8:143–6.

[20] Fisher DA, Rapp GF, Emkes M. Idiopathic scoliosis: transcutaneous muscle stimulation versus the Milwaukee brace. Spine 1987;12: 792–801.

[21] Galante J, Schultz AB, deWald RL, Ray RD. Forces acting in the Milwaukee brace in patients under treatment for idiopathic scoliosis. J Bone Joint Surg Am 1970;52:498–506.

[22] Green NE. Part-time bracing of adolescent idiopathic scoliosis. J Bone Joint Surg Am 1986;68:738–42.

[23] Montgomery F, Willner S. Prognosis of brace-treated scoliosis. Comparison of the Boston and Milwaukee methods in 244 girls. Acta Orthop Scand 1989;60:383–5.

[24] Veldhuizen AG, Cheung J, Bulthuis G, Nijenbanning G. A new orthotic device in the non-operative treatment of idiopathic scoliosis. Med Eng Phys 2002;24:209–18.

[25] Rigo M, Negrini S, Weiss HR, et al. SOSORT consensus paper on brace action: TLSO biomechanics of correction. Scoliosis 2006;20:1–11.

[26] Udén A, Willner S. The effect of lumbar flexion and Boston thoracic brace on the curves in idiopathic scoliosis. Spine 1983;8:846–50.

[27] Willner S. Effect of the Boston thoracic brace on the frontal and sagittal curves of the spine. Acta Orthop Scand 1984;55:457–60.